MRI ANTENNA

Benefit of Provisional Application

The present application claims the benefit of U.S.S.N. 60/172,199, filed on December 17, 1999, assigned to the assignee of the present invention and incorporated by reference, herein.

Related Applications

The present application is related to U.S.S.N. __________ (Attorney Docket No. 5007-4048) and U.S.S.N. __________ (Attorney Docket No. 5007-4049), both filed on the same day as the present invention, assigned to the assignee of the present invention and incorporated by reference, herein.

Field of the Invention

This invention relates to radio frequency receiving and transmitting antennas, and, more particularly, to receiving and transmitting radio frequency antennas for use in magnetic resonance imaging.

Background of the Invention

Magnetic resonance imaging ("MRI") is a well known, highly useful technique for diagnosing abnormalities in biological tissue. MRI can detect abnormalities which are difficult or impossible to detect by other techniques, without the use of x-rays or invasive procedures.

MRI uses changes in the angular momentum or spin of the atomic nuclei of certain elements within body tissue in a static magnetic field after excitation by radio frequency energy, to derive images containing useful information concerning the condition of the tissue. During an MRI procedure, the patient is inserted into an imaging volume containing a static magnetic field. The vector of the angular momentum or spin of nuclei containing an odd number of protons or neutrons tends to align with the direction of the magnetic field. A transmitting antenna within the imaging volume emits a pulse or pulses of radio frequency energy having a particular bandwidth of frequency, referred to as the resonant or Larmor frequency, shifting the vectors of the nuclei out of alignment with the applied magnetic field. The spins of the nuclei then turn or "precess" around the direction of the applied primary magnetic field. As their spins precess, the nuclei emit small radio frequency signals, referred to as magnetic resonance ("MR") signals, at the resonant or Larmor frequency, which are detected by a radio frequency receiving antenna tuned to that frequency. The receiving antenna is typically positioned within the imaging volume proximate the patient. Gradient magnetic fields are provided to spatially encode the MR signals emitted by the nuclei. After the cessation of the application of radio frequency waves, the precessing spins gradually drift out of phase with one another, back into alignment with the direction of the applied magnetic field. This causes the MR signals emitted by the nuclei to decay. The MR signals detected by the receiving antenna are amplified, digitized and processed by the MRI system. The same antenna may act as the transmitting and receiving antenna. Hydrogen, nitrogen-14, phosphorous-31, carbon-13 and sodium-23 are typical nuclei detected by MRI. Hydrogen is most commonly detected because it is the most abundant nuclei in the human body and emits the strongest MR signal.

The rate of decay of the MR signals varies for different types of tissue, including injured or diseased tissue, such as cancerous tissue. By known mathematical techniques involving correlation of the gradient magnetic fields and the particular frequency of the radio frequency waves applied at various times with the rate of decay of the MR signals emitted by the patient, it is possible to determine the concentrations and the condition of the environment of the nuclei of interest at various locations within the patient's body. This information is typically displayed as an image with varying intensities, which are a function of the concentration and environment of the nuclei of interest. Typical MRI systems are the Quad 7000 and Quad 12000 available from FONAR Corporation, Melville, New York, for example.

The quality of the magnetic resonance image is directly related to the characteristics of the receiving and transmitting antenna. Significant electrical characteristics of the antenna include its sensitivity, Q factor and the signal-to-noise ratio.

Sensitivity is the signal voltage generated in the receiving antenna by MR signals of a particular magnitude. The higher the sensitivity within the region to be imaged, the weaker the signals which can be detected. The sensitivity of the antenna is preferably substantially uniform with respect to MR signals emanating from all volume elements within the region of the subject which is to be imaged.

The Q or quality factor, which is closely related to the sensitivity of the antenna, is a measure of the ability of the antenna to amplify the received signal. The Q-value of the antenna can be lowered by a patient proximate or within an antenna, due to capacitive and to a lessor extent the inductive coupling between the patient and the antenna. Antennas must therefore have a high Q-value when they are unloaded and the Q-value must not become too

diminished by the presence of the patient. On the other hand, the coil must couple well with the region of a patient's anatomy which is to be imaged.

Signal-to-noise ("S/N") ratio is the ratio between those components in the electrical impulses appearing at the antenna terminals representing the detected MR signals and the components representing spurious electromagnetic signals in the environment and internally generated thermal noise from the patient. To optimize the S/N ratio, the antenna should have low sensitivity to signals from outside the region to be imaged and to thermal noise. To enhance both S/N ratio and sensitivity, the antenna is "tuned" or arranged to resonate electrically at the frequency of the MR signals to be received (the Larmor frequency), which is typically several megahertz or more. Neither the coil size nor geometry of the antenna can be allowed to create an inductance or self-capacitance which prevents tuning to the desired frequency.

The antenna must also meet certain physical requirements. The antenna should have a high filling factor, which maximizes the amount of tissue which fits within the volume detected by the windings of the coil. The antenna must also fit within the relatively small imaging volumes typically provided for receiving the subject within the magnet assembly, along with other components of the system and the subject. The antenna should not cause significant discomfort to the subject. Additionally, the antenna should be easy to position with respect to the subject, and be relatively insensitive to minor faults in positioning relative to the subject.

These numerous considerations often conflict with one another and therefore must be balanced during the design process.

The sensitivity and S/N ratio of MRI radio frequency receiving antennas have been improved by positioning a first coil, tuned to resonate at the Larmor frequency of the species of interest, proximate the part of the subject which is to be imaged, and positioning a

similarly tuned second coil, typically a single loop, adjacent to the first coil. The second coil is connected to the preamplifier of the MRI system. The first and second coils are inductively coupled to each other. MR signals emitted by the patient induce voltages in the first winding, causing current to flow within the winding. The current generates a magnetic field which induces voltage in the second winding. The MR signals may induce voltages in the second coil, as well. The voltages induced in the second coil are processed by the MRI system. Use of such first and second coils amplify the MR signals, and the second coil filters spurious signals outside of the frequency band of the Larmor frequency. See, for example, U.S. Patent No. 5,583,438 and U.S. Patent No. 5,575,287, assigned to the assignee of the present invention.

Radio frequency antenna coils may be used in a variety of configurations. For example, the coil may be receiving coil, as discussed above. The receiving coil may be part of an array of receiving coils, such as in the primary and secondary coil arrangements, discussed above. The receiving coil may also act as the transmitting coil of the MRI system. A pair of receiving coils can also be arranged 90^{0} with respect to each other to enable quadrature detection, which improve the signal-to-noise ratio.

SUMMARY OF THE INVENTION

In accordance with the present invention, receiving and transmitting antennas for use in magnetic resource imaging comprise inductively coupled first and second windings tuned to the same frequency, typically the Larmor frequency of species of interest. In a receiving antenna, the second winding is connected to the receiver subsystem of an MRI system.

It is believed, without limiting the scope of the invention, that in antennas of the design of the present invention, the first winding substantially shields the second winding from direct reception of magnetic resonance ("MR") signals. However, since the windings are

inductively coupled, MR signals detected by the first winding induce voltage signals in the second winding. Since both windings are tuned to the same frequency, highly filtered signals are provided for analysis by the MRI system. The first and second windings may form a coaxial cable. Multiple coaxial cables may be suitably coupled to form antenna arrays for use with different parts of the body.

In a transmitting antenna, the second winding is connected to the radio frequency power source in the radio frequency transmitting section of the MRI system. Voltage signals provided to the second winding induce voltage signals and current flow in the first winding, causing the emission of highly filtered radio frequency signals.

In accordance with one embodiment of the present invention, an MRI antenna comprises an inner conductor with first and second ends for being electrically connected across a capacitor to tune the inner conductor to a frequency. The first and second ends provide an output of the antenna. An outer conductor substantially surrounds the inner conductor. The outer conductor also has first and second ends for being electrically connected across a capacitor to tune the outer conductor to the same frequency as the inner conductor. The inner and outer conductors define a region for receiving a body part and are inductively coupled during operation.

The inner and outer conductors preferably define a coaxial cable unit. Multiple coaxial cable units may be provided, wherein the inner conductors of each unit are connected to form a first circuit tunable to a frequency. The outer conductors are also connected to form a second circuit tunable to the frequency. The output is provided from the first circuit. Multiple inner conductors may also be provided, electrically connected to form a circuit tunable to the frequency.

A second outer conductor may be also provided around the first outer conductor, with holes therethrough. It is believed, without limiting the scope of the invention, that magnetic resonance signals may be detected by the first outer conductor, due to the presence of the holes, and that magnetic resonance signals detected by the second outer conductor also induce signals in the first outer conductor. The ends of the second outer conductor are connected through a capacitor to tune the outer conductor to the frequency, as well.

In accordance with another embodiment of the invention, an MRI antenna comprises a first winding and a second winding. The first and second windings are inductively coupled and are each tuned to the same frequency. The first winding substantially shields the second winding from direct detection of magnetic resonance signals emitted from a subject and the second winding provides an output of the antenna.

In accordance with another embodiment of the invention, an MRI antenna comprises a coaxial cable comprising an inner conductor and an outer conductor surrounding the inner conductor. The outer conductor and the inner conductor are inductively coupled and tuned to the Larmor frequency of the species of interest during operation. The inner conductor provides an output of the antenna.

In accordance with another embodiment of the invention, an MRI antenna comprises first, second and third coaxial cable units, each having an outer conductor and an inner conductor. The outer conductor of each unit substantially surrounds the respective inner conductor. The outer conductor and the inner conductors of each unit are inductively coupled to each other during operation. Adjacent coaxial cable units are also inductively coupled during operation. Each coaxial cable unit defines a region for receiving a body part. The outer conductors are electrically connected to form a first circuit tunable to a frequency and the inner

conductors are electrically connected to form a second circuit tunable to the frequency. An output from the antenna is provided from the second circuit of the inner conductors. The first, second and third coaxial cable units may lie in parallel planes and may also be aligned along an axis perpendicular to the first, second and third parallel planes. Such an antenna is particularly suited for magnetic resonance imaging of a hand, wrist or toe.

In accordance with another embodiment of the invention, an MRI antenna comprises a first inner conductor with first and second ends and a first outer conductor substantially surrounding the first inner conductor. The first inner conductor and the first outer conductor define a first coaxial cable unit defining a region for receiving a body part. A second inner conductor and a second outer conductor substantially surrounding the second inner conductor, defining a second coaxial cable unit, are also provided. The second inner conductor and the second outer conductor are inductively coupled during operation. The first and second coaxial cable units are concentric and lie in substantially the same plane. The second coaxial cable unit is within a region defined by the first coaxial cable unit. The inner conductors of the first and second coaxial cable units are connected in parallel across a capacitor. The outer conductors are electrically connected as well. Ends of the first coaxial cable unit are also electrically connected across a capacitor. The circuits comprising the inner and outer conductors are tunable to the same frequency. An output of the antenna is provided from the inner conductors.

In accordance with another embodiment of the invention, an MRI antenna comprises a support and first and second coaxial cable units supported by the support in a first plane to define a region for receiving a body part. Each coaxial cable unit comprises an inner conductor and an outer conductor substantially surrounding the inner conductor. The inner

conductors of the first and second coaxial cable units are electrically connected, preferably through capacitors, to form a tunable circuit. The outer conductors of the first and second coaxial cable units are also electrically connected, preferably through capacitors, to form a second circuit tunable to the frequency. An output of the antenna is provided from the circuit including the inner conductors.

Third and fourth coaxial cable units may further be supported by the support adjacent to the first and second coaxial cable units in a second plane, and fifth and sixth coaxial cable units may be supported by the support adjacent to the first and second coaxial cable units in a third plane. The third plane is on an opposite side of the first and second coaxial cable units as the second plane. The inner and outer conductors of the third, fourth, fifth and sixth coaxial cable units are connected in series, preferably through capacitors, to form a single circuit tunable to the frequency. The inner and outer conductors are inductively coupled during operation.

Seventh and eighth coaxial cable units may also be supported by the support in a fourth plane between the first plane and the third plane to define a further region for receiving the body part. The seventh and eighth coaxial cable units also comprise an inner conductor and an outer conductor substantially surrounding the inner conductor, and are inductively coupled during operation. The inner conductors are electrically connected to form a circuit tunable to the frequency and the outer conductors are preferably electrically connected to form a circuit tunable to the frequency, also preferably, through capacitors.

Preferably, ends of the inner conductors and ends of the outer conductors in the pairs of coaxial cable units in each plane are electrically connected through capacitors, to lower the inductance of the respective circuits. Such an antenna is suitable for imaging the head and neck.

Magnetic resonance imaging systems comprising such antennas are also disclosed. The antennas may also be used as transmitting antennas.

BRIEF DESCRIPTION OF THE DRAWINGS

Fig. 1 is a plan view of a basic coaxial cable unit 10 of an MRI antenna in accordance with one embodiment of the present invention;

Fig. 2 is a cross-sectional view of the basic coaxial cable unit 10, along line 2-2 in Fig. 1;

Fig. 3 shows the basic coaxial cable unit of Fig. 1, with electrical connections to form an antenna in accordance with one embodiment of the present invention;

Fig. 4 is a schematic diagram of a circuit corresponding to the antenna of Fig. 3;

Fig. 5 shows the basic unit of Fig. 1, with a shielding adapter;

Fig. 6 shows another shielding technique;

Fig. 7 shows a variation of the basic unit of Fig. 1, comprising two sections of coaxial cable, and their electrical connections;

Fig. 8 is a schematic diagram of a circuit corresponding to the configuration of Fig. 7;

Fig. 9 is a schematic representation of an antenna array 100 appropriate for imaging of a hand, wrist or toe, using three basic coaxial cable units, in accordance with the present invention;

Fig. 10 is a schematic representation of the side view of the antenna array of Fig. 9;

Fig. 11 is a schematic representation of a circuit corresponding to the antenna array of Fig. 9;

Fig. 12 is a perspective view of the antenna array of Fig. 9, encased in a base of dielectric material, in use with a patient;

Fig. 13 is a schematic diagram of an antenna array in accordance with another embodiment the present invention, which is particularly suited for imaging the head and neck;

Fig. 14 is a side view of the antenna array of Fig. 13, wherein four coaxial units are each in respective vertical, parallel planes, and the center of each unit lies along the same axis;

Fig. 15 is a schematic diagram of a circuit corresponding to the antenna array of Fig. 11;

Figs. 16-18 show coaxial cable units in various test configurations;

Fig. 19 shows a test configuration of a coaxial cable unit of the present invention;

Fig. 20 shows an antenna array comprising two concentric coaxial cable units lying in the same plane;

Fig. 21 is a schematic diagram of a circuit corresponding to the configuration of Fig. 20;

Fig. 22 shows a quadrature antenna system comprising the hand, wrist and toe antenna array of Figs. 9-12 and a pair of rectangular coaxial cable units;

Fig. 23 is a top view of the quadrature antenna system of Fig. 22, showing the rectangular coaxial cable units;

Fig. 24 shows a quadrature antenna system of Fig. 23, including additional coaxial cable units coupled to the rectangular coaxial cable units;

Fig. 25 is a side view of the quadrature antenna system of Fig. 24, along line 25-25 in Fig. 24;

Fig. 26 is a side view of the quadrature antenna system of Fig. 24, along line 26-26 of Fig. 25;

Fig. 27 is a schematic diagram of the preamplifier section of an MRI system for use with the quadrature antenna system of Figs. 22-26;

Fig. 28 is a side view of the gap between the magnet poles of an MRI system;

Fig. 29 is a plan view of a coaxial cable unit in accordance with another embodiment of the invention, including four inner conductors;

Fig. 29a is a cross-sectional view of the coaxial cable unit of Fig. 29, along line 29-29;

Fig. 30 shows a preferred connection scheme for the inner conductors of the configuration of Fig. 29;

Fig. 31 shows three coaxial cable units as in Fig. 29, separated by an insulator;

Fig. 32 shows the three coaxial cable units of Fig. 31 and preferred connections between their inner and outer conductors;

Fig. 33 is a schematic diagram of a circuit corresponding to the antenna array of Fig. 31;

Fig. 34 shows a triaxial cable unit in accordance with another embodiment of the present invention;

Fig. 34 is a cross-sectional view of the triaxial cable unit of Fig. 34, along line 34a-34a;

Fig. 35 is a schematic representation of the triaxial cable unit of Fig. 34;

Fig. 36 is a schematic representation of the triaxial cable unit of Fig. 34, wherein the inner conductors are connected in parallel;

Fig. 37 shows a coaxial cable unit as in Fig. 3, wherein the ends of the inner conductor are connected across a capacitor to a radio frequency power source;

Fig. 38 is a schematic representation of the coaxial cable unit of Fig. 37; and

Fig. 39 is a schematic representation of portions of an MRI system, showing in particular a connection between an antenna or antenna array 802 in accordance with the present invention with certain components of the MRI system;

DETAILED DESCRIPTION OF THE INVENTION

The unit 10 comprises an inner conductor 12 and an outer conductor 14 coaxially arranged, as shown in the cross-sectional view of Fig. 2. The inner conductor 12 and the outer conductor 14 are separated by a dielectric material 24, such as Teflon®, for example, forming a coaxial cable unit 10. The coaxial conductors 12, 14 are inductively coupled to each other. Preferably, the inner and outer conductors 12, 14 are tightly coupled to each other. More preferably, they are critically coupled to each other.

The inner conductor 12 has two ends 16, 18 and the outer conductor 14 has two ends 20, 22. A plurality of basic units 10 may be connected or coupled to each other in different combinations to form an antenna array, as discussed below.

A body part to be imaged is received in the region 26 bounded by the coaxial cable unit 10. In Fig. 1, the coaxial conductors are in the shape of a ring. Other shapes may be used, dependent on the body part being imaged.

In one example, the outputs 16, 18 of the inner conductor 12 are connected to each other through a capacitor C_1 and the outputs 20, 22 of the outer conductor 14 are connected to each other through a capacitor C_2 to form an antenna 11, as shown in Fig. 3. The capacitors C_1, C_2 have values such that the inner and outer conductors 12, 14 are tuned to the Larmor

frequency of the species of interest, such as hydrogen. The capacitor C_1 may be connected to a variable capacitor ("varactor") in the preamplifier of the receiver system of an MRI system, or it may be the varactor in the preamplifier. Each conductor 12, 14 may be copper tubing. As shown in Fig. 2, the inner diameter "d_1" of the inner conductor 12 may be about 0.125 inches. The diameter "d_2" of the outer conductor 14 may be about 0.25 inches. The wall thickness of each conductor 12, 14 may be about 0.032 inches.

The coaxial cable unit 10 may be made from readily available soft copper refrigeration tubing of appropriate diameters. Such tubing may be obtained from Metal Product, Wynne, Arkansas, for example. The tubing corresponding to the inner conductor 12 is covered by a Teflon® tubing, such as TFT70C polytetrafluoroethylene, available from AIN Plastics, Inc., Mount Vernon, NY, for example. The Teflon® covered inner conductor 12 is inserted through the tubing corresponding to the outer conductor 14, forming the coaxial cable unit 10. The coaxial cable unit 10 may also be flexible. One-quarter inch, High Power, High Temperature Dielectric Coaxial Cable, Andrew HST1-50 HELIAX, available from Andrew Corporation, Orland Park, Illinois, may be used, for example.

The inner conductor 12 is shielded from direct reception of the MR signals by the outer conductor 12. However, voltage signals induced in the outer conductor 14 induce voltage signals in the inner conductor 12. The outer conductor 14 may, therefore, be modeled as a primary winding of a transformer while the inner conductor 12 may be modeled as the secondary winding of the transformer.

Fig. 4 is a schematic diagram of a circuit corresponding to the antenna of Fig. 3. The inner conductor 12 and the outer conductor 14 are represented as inductors L_1, L_2, respectively. The capacitors C_1 and C_2 are shown, as well. The inductor L_1 and the capacitor C_1

form a first circuit "A" and the inductor L_2 and the capacitor C_2 form a second circuit "B". The inductors L_1, L_2 are inductively coupled to each other and have a distributed capacitance Cd. The two circuits are tuned to the same, Larmor frequency, forming anti-resonant or parallel resonant, double-tuned circuits. The circuits A, B form a bandpass network which filters frequencies outside of the bandpass of the two circuits. The bandpass of the circuit B (corresponding to the inner conductor 12) is narrower than the bandpass of the circuit A (corresponding to the outer conductor 14), as is known in the art. Noise and other signals outside of the narrower bandwidth of the inner conductor 12 are therefore filtered. As mentioned above, in the embodiment of Fig. 3, the inner conductor 12 is connected to the receiver subsystem of an MRI device, providing highly filtered signals for analysis.

The outer conductor 14 also shields the inner conductor 12 from capacitive coupling with the body of the patient over most of its length. As shown in Fig. 3, however, the inner conductor 12 may not be effectively shielded by the outer conductor 14 in the gap 28 between the ends of the outer conductor 14. To improve the shielding effect of the outer conductor 14 on the inner conductor 12, the ends 20, 22 of the outer conductor 14 may be connected through a copper tube adaptor 30, which closes the gap 28 between the ends of the inner and outer conductors 12, 14, as shown in Fig. 5. The adaptor 30 acts as a capacitive connection between the ends 20, 22 of the outer conductor 14, as well as shielding the outputs 16, 18 of the inner conductor 12. The adaptor 30 also minimizes field distributions in the gap 28 between the ends of the conductors 12, 14.

Two capacitors C_1, C_2 are provided within the adaptor 30, electrically connecting in series the first and second ends 16, 18 of the inner conductor 12 and the first and second ends

20, 22 of the outer conductor 14, respectively. The exposed ends, 16, 18 of the inner conductor 14 and the capacitors C_1, C_2 are shielded by the adaptor 30.

The direct exposure of the inner conductor 12 to MR signals from within the area 28 may also be minimized by distancing the exposed ends 16, 18 from the source of the MR signals in the region 28 and positioning more of the outer conductor 14 between the inner conductor 12 and the source of MR signals within the coil, as shown in Fig. 6.

Dependent on the overall length of the coaxial ring and the magnetic field strength of the magnet used in the MRI system, in order to tune the antenna 11 of Fig. 3 to the Larmor frequency, the inductance of the inner and outer conductors 12, 14 of the coaxial cable may need to be decreased. The inductance may be decreased by decreasing the lengths of the inner and outer conductors 12, 14, as is known in the art. For example, the inner and outer conductors 12, 14 may be split into two sections 34, 36, as shown in Fig. 7, to facilitate their being tuned to the higher frequencies used in higher field strength magnets. The ends of the inner conductor 12 of a first section 34 are connected in series through respective capacitors C_1, C_3 to the ends of the inner conductor 12" of the second section 36. The ends of the outer conductor 14" of the first section 34 are similarly connected in series to the ends of the outer conductors 14" of the second section 36 through respective capacitors C_2, C_4. The values of the capacitors C_1, C_2, C_3, C_4 are adjusted to tune the conductors 12', 12", 14', 14" to the desired frequency. Either the capacitor C_1 or the capacitor C_3 may be connected to the variable capacitor in the preamplifier of the MRI system, or may be the variable capacitor in the preamplifier.

Fig. 8 is a schematic diagram of the circuit corresponding to the configuration of Fig. 7. In this case, the inner conductors 12', 12" correspond to the inductors L_1, L_2 which are connected in series through capacitors C_1, C_3. The outer conductors 14', 14" correspond to the

inductors L_3, L_4, respectively, which are connected in series through the capacitors C_2 and C_4. As above, the outer conductor 14 (L_3, L_4) and the inner conductor 12 (L_1, L_2) are inductively coupled.

Fig. 9 is a schematic diagram of an antenna array 100 appropriate for imaging of a hand, wrist or toe, using three coaxial cable ring units 102, 104, 106 as in Fig. 1, in accordance with one embodiment of the present invention. The coaxial ring units 102, 104, 106 lie in three respective parallel, vertical planes P1, P2, P3, respectively, as shown in Fig. 10. To better accommodate the hand, wrist or toe, such that the outer conductors are close to the body party, each unit has a rectangular shape. The antenna array 100 may be used for imaging other body parts, as well. The shape of each unit may be varied as appropriate to accommodate and maintain suitable proximity to the body part.

Each unit 102, 104, 106 has an inner conductor 12, 12', 12", respectively, and an outer conductor 14, 14', 14". A first, top end 16 of the inner conductor 12 of the first unit 102 is electrically connected in series to a first, bottom end 16' of the inner conductor 12' of the second unit 104 through a capacitor C_1. The second, top end 18' of the inner conductor 12' is electrically connected in series to the bottom end 16" of the inner conductor 12" of the third unit 106 through a capacitor C_2. The second, bottom end 18 of the inner conductor 12 of the first unit 102 and the second, top end 18" of the inner conductor 12" of the third unit 106 provide an output 108 of the antenna array 100. The output 108 may be connected across a capacitor (not shown) which may be connected to or may be the varactor of the preamplifier of the receiver system of the MRI device. The output 108 may be connected to the preamplifier through a BNC connector acting as a capacitive module.

The top end 20 of the outer conductor 14 of the first unit 102 is electrically connected in series to the bottom end 20' of the outer conductor 14' of the second unit 104, through a capacitor C_3. The top end 22' of the outer conductor 14' is electrically connected in series to the first, bottom end 20" of the outer conductor 14" of the third unit 106 through a capacitor C_4. The bottom end 22 of the outer conductor 14 of the first unit 102 and the second, top end 22" of the third unit 106 are electrically connected in series through the capacitor C_5, which is part of a BNC connector acting as a capacitive module.

The antenna array 100 of Fig. 9 may be represented as three-double tuned transformers connected in series, as shown in Fig. 11. The inductors L_1, L_2, L_3 correspond to the inner conductors 12, 12', 12" of the first, second and third units 102, 104, 106, respectively, and the inductors L_4, L_5, L_6 correspond to the outer conductor 14, 14' and 14". In one implementation, for use in an MRI system including a magnet with a field strength of from about 6,000 Gauss to about 8,000 Gauss, capacitor C_1=68 picofarads; capacitor C_2=56 picofarads; capacitor C_3=22 picofarads; and capacitor C_4=22 picofarads. The inner and outer conductors 12, 14 were tuned to the Larmar frequency of hydrogen for the particular magnet field strength, which in the range of 6,000 to 8,000 Gauss is between about 25-35 megahertz. The length "D1" (Fig. 9) of each coaxial cable unit 102, 104, 106 was 7 inches. The distance "D_2" (Fig. 10) between the center of adjacent coaxial cable units was 1.5 inches. The Q of the antenna array 100 was found to be about 197 while the S/N ratio was found to be 320 to 1.

While an antenna array comprising three coaxial cable units preferred, the antenna array 100 could comprise the two coaxial cables 102 and 106, in which case, the first end 16 of the inner conductor 12" of the first coaxial cable 102 would be connected to the first end 16 of the inner conductor 12 coaxial cable unit 106 and the first end 20 of the outer conductor 14

would be connected to the first end 20" of the outer conductor 14" coaxial cable unit 106. The outputs of the antenna array would be the same.

The antenna array 100 is preferably encased in a base 100a of a rigid, dielectric material, such as the fire resistant polymers polyvinyl chloride, polytetrafluoroethylene and fluorinated ethylene propylene, as shown in Fig. 12. Polymers having low dielectric constants are particularly preferred. In Fig. 12, the first, second and third coaxial cable units 102, 104, 106 are shown in phantom. The coaxial cable units 102, 104, 106 are coaxially aligned along an axis A through the center of the base 100a, as discussed above with respect to Fig. 11. The base 100a is connected to a plate 100b which may be positioned on the patient bed within the gap of the MRI magnet. A patient's hand ("h") is shown, partially inserted into a region within each of the coaxial cable units 102, 104, 106, for magnetic resonance imaging of the fingers, for example. A strap 100c may be provided to restrain movement of the hand h during imaging.

Flexible coaxial cable units can also be supported by a flexible belt. An example of a flexible coaxial cable is described above. The flexible belt can comprise cross-linked polyethylene foam, available from Contour Fabricators, Inc., Grand Blanc, Michigan.

Fig. 13 is a schematic diagram of an antenna array 200 in accordance with another embodiment the present invention, which is particularly suited for imaging the head and neck. Fig. 14 is a side view of the array, wherein four coaxial ring units 202, 204, 206, 208 are each in respective vertical, parallel planes P1, P2, P3, P4 and the center of each unit lies along the same axis "A".

Returning to Fig. 13, because larger conductors are needed to surround the head, each coaxial ring unit preferably has two sections to lower the inductance of the conductor, as

discussed above with respect to the embodiment of Fig. 7. Each section has an outer conductor with first and second ends and an inner conductor with first and second ends.

In the first unit 202, the upper section 202a has an inner conductor 210 with first and second ends 210a, 210b and an outer conductor 214 with first and second ends 214a, 214b. Similarly, the lower section 202b has an inner conductor 216 with first and second ends 216a, 216b and an outer conductor 218 with first and second ends 218a, 218b. The first end 210a of the inner conductor 210 is electrically connected in series to the first end 216a of the inner conductor 216 through a capacitor C_1. The first end 214a of the outer conductor 214 is electrically connected in series to the first end 218a of the outer conductor 218 through a capacitor C_2. The second end 210b of the inner conductor 210 and the second end 216b of the inner conductor 216 provide an Output. The Output would be connected to a varactor in the preamplifier of the receiver system of the MRI device, optionally through a capacitor. The second end 214b of the outer conductor 214 is electrically connected in series to the second end 218b of the outer conductor 218 through a capacitor C_3 of a BNC connector capacitive module.

In the second unit 204, the upper section 204a has an inner conductor 220 with first and second ends 220a, 220b and an outer conductor 222 with first and second ends 222a, 222b. Similarly, the lower section 204b has an inner conductor 224 with first and second ends 224a, 224b and an outer conductor 226 with first and second ends 226a, 226b. The first end 220a of the inner conductor 220 is electrically connected in series to the first end 224a of the inner conductor 224 through a capacitor C_4. The first end 222a of the outer conductor 222 is electrically connected in series to the first end 226a of the outer conductor 226 through a capacitor C_5. The second end 220b of the inner conductor 220 and the second end 224b of the inner conductor 224 are electrically connected in series through a capacitor C_{14}. The second end

222b of the outer conductor 222 is electrically connected in series to the second end 226b of the outer conductor 226 through a capacitor C_6.

In the third unit 206, the upper section 206a has an inner conductor 228 with first and second ends 228a, 228b and an outer conductor 230 with first and second ends 230a, 230b. Similarly, the lower section 206b has an inner conductor 232 with first and second ends 232a, 232b and an outer conductor 234 with first and second ends 234a, 234b. In the fourth unit 208, the upper section 208a has an inner conductor 236 with first and second ends 236a, 236b and an outer conductor 238 with first and second ends 238a, 238b. The lower section has an inner conductor 240 with first and second ends 240a, 240b and an outer conductor 242 with first and second ends 242a, 242b.

The first end 228a of the inner conductor 228 of the third unit 206 is electrically connected in series to the first end 232a of the inner conductor 232 through a capacitor C_7. The first end 230a of the outer conductor 230 is electrically connected in series to the first end 234a of the outer conductor 234 through a capacitor C_8. The second end 230b of the outer conductor 230 is electrically connected in series to the second end 232b of the inner conductor 232 through a capacitor C_9. The second end 228b of the inner conductor 228 is electrically connected in series to the second end 242b of the outer conductor 242 of the lower section 208b of the fourth unit 208 through a capacitor C_{10}. The second end 234b of the outer conductor 234 is electrically connected in series to the second end 236b of the inner conductor 236 of the upper section 208a of the fourth unit 208 through a capacitor C_{11}. The second end 238b of the outer conductor 238 is electrically connected in series to the second end 240b of the inner conductor 240 through a capacitor C_{12}. The first end 236b of the inner conductor 236 is electrically connected in series to the first end 240a of the inner conductor 240 through a capacitor C_{13}. The first end 238a of the

outer conductor 238 is electrically connected in series to the first end 242a of the outer conductor 242 through a capacitor C_{14}.

In operation, a magnetic field is generated between the third unit 206 and the fourth unit 208. The fourth unit 208, which has a smaller diameter than the third unit 206, reflects the magnetic field towards the third unit 206. The first and second units 202, 204 act as receiver coils and resonators within the magnetic field created by the third and fourth units.

Fig. 15 is a corresponding circuit diagram of the coaxial cable units of Fig. 11. The inductors of Fig. 13 correspond to the inner and outer conductors of Fig. 11, as follows:

L_1	Inner Conductor 210	L_9	Inner Conductor 228
L_2	Inner Conductor 216	L_{10}	Inner Conductor 232
L_3	Outer Conductor 214	L_{11}	Outer Conductor 230
L_4	Outer Conductor 218	L_{12}	Outer Conductor 234
L_5	Inner Conductor 220	L_{13}	Inner Conductor 236
L_6	Inner Conductor 224	L_{14}	Inner Conductor 240
L_7	Outer Conductor 222	L_{15}	Outer Conductor 238
L_8	Outer Conductor 226	L_{16}	Outer Conductor 242

The corresponding capacitors in Figs. 11 and 13 are commonly identified.

In one configuration of the head and neck antenna array 200 for the use in an MRI system with a magnet with a field strength of between about 6,000 Gauss to about 8,000 Gauss, the capacitors C_1, C_2, C_4, C_9, C_{11} and C_{14} had a value of 220 picofarads ("pf"). Capacitor C_3 had a value of 470 pf and Capacitor C_{10} had a value of 820 pf. The inner and outer conductors are preferably tuned to a frequency between about 25 to about 35 MHz. The first, second and third coaxial cable units 202, 204, 206 had an outer height "h1" (Fig. 13, Fig. 14) of about 11 inches.

The fourth coaxial ring unit 208 had an outer diameter "D2" of about 9 inches. The distance "D" between adjacent coaxial cable units was 2 inches. (Fig. 14) Each coaxial cable unit was critically coupled to an adjacent coaxial cable unit. Such an antenna array 200 was found to have a Q of about 220 and a S/N ratio of about 360 to 1.

When positioned over the head of a patient, the third unit 204 is preferably positioned over the eyes of the patient.

The antenna array 200 may be encased in a base 200a of rigid dielectric material, as discussed above with respect to Fig. 13.

Testing of several of the configurations discussed above show significant improvements in Q and S/N ratio. The Q of the antenna may be measured by a network analyzer, such as the 3577A, available from the Hewlett Packard Company. The S/N ratio may be measured by performing a phantom scan by an MRI system, as is known in the art. In one test, the ends 20, 22 of an outer conductor 14 were electrically connected in series across a variable capacitor C_1, as shown in Fig. 16. The range of the variable capacitor C_1 was from 0 to about 30 picofarads. The outer conductor was tuned to a frequency between 25 to 35 MHz. The Q and S/N ratio of the coaxial cable in Fig. 16 were found to be 59 and 180, respectively. In Fig. 17, the ends 16, 18 of the inner conductor 12 were connected across the variable capacitor C_1, and the ends 20, 22 of the outer conductors were connected across a capacitor C_2 having a value of 10 picofarads. The inner and outer conductors 12, 14 were tuned to the frequency of between about 25 to 35 MHz. The Q of the circuit was 72 and the S/N ratio was 220, approximately a 22% improvement in both characteristics. The use of multiple coaxial cable units has been found to further improve the Q and S/N ratio of the antenna. As mentioned above, the Q of the hand,

wrist and toe antenna array 100 of Fig. 10, for example, was found to be about 197, about a 233% improvement. The S/N ratio was found to be about 320, about an 82% improvement.

When only the inner ends 16, 18 of the conductor 12 of a coaxial cable unit were connected across the capacitor C_1 and a capacitor C_3 of 120 picofarads, and the inner conductor was tuned to a frequency between about 25 to 35 MHz, as in Fig. 18, the Q of the circuit 28 was 103. The S/N ratio was 160. In the configuration of Fig. 19, the ends 16, 18 of the inner conductor 12 and the ends 20, 22 of the outer conductor 14 were connected through capacitors C_1, C_4, C_5 and C_6, respectively. The capacitors C_4 = 72 pf, the capacitor C_5 = 68 pf and the capacitor C_6 = 47 pf. Both conductors were tuned to a frequency between about 25 to 35 MHz. The Q of the circuit was 128 and the S/N ratio was 200 to 1, approximately a 25% improvement. As mentioned above, the use of multiple coaxial cables further improves the Q and S/N of the antenna. The Q of the head antenna array 200, with four coaxial cables, was 220, about a 113% improvement over the configuration of Fig. 18. The S/N ratio was 360 to 1, about a 125% improvement.

The basic coaxial cable unit 10 of Fig. 1 may also be used in antennas in other configurations, in accordance with the present invention. For example, in Fig. 20, an antenna array 400 comprises two concentric coaxial cable units 402 and 404, lying in the same plane. The inner coaxial cable unit 402 comprises an inner conductor 406 and an outer conductor 408. The inner conductor 406 has a first end 414 and a second end 418. The outer coaxial cable unit 404 comprises an inner conductor 410 and an outer conductor 412. The inner conductor 410 has a first end 416 and a second end 420. The outer conductor 412 has a first end 422 and a second end 424.

The first end 414 and the second end 418 of the inner conductor 406 of the inner coaxial cable unit 402 are electrically connected in parallel to the first end 416 and the second end 420, respectively, of the inner conductor 410 of the outer coaxial cable 404. A capacitor C_1 is electrically connected in parallel to the inner conductors 406 and 410.

The first end 422 of the outer conductor 412 is electrically connected to the second end 424 of the outer conductor 412, through a capacitor C_2. A portion 408a of the outer conductor 408 of the inner coaxial cable unit 402 is also directly electrically connected to an adjacent portion 412a of the outer conductor 412 and a portion 408b of the outer conductor 408 is directly electrically connected to a portion 412b of the outer conductor 412b, through electrical contacts 411, 413, respectively. Insulation (not shown) may be provided between the outer conductors 408 and 412 along the remainder of their lengths.

The inner conductors 406, 410 of the two coaxial cables 402, 404 provide the output of the antenna 400. The capacitor C_1 may be connected to a variable capacitor in the preamplifier of the MRI system, or be the variable capacitor of the preamplifier.

As above, the values of the capacitors C_1 and C_2 are such that the circuit including the inner conductors and the circuit including the outer conductors, are tuned to the same Larmor frequency.

Fig. 21 is an schematic diagram of a circuit corresponding to the configuration of Fig. 20, wherein the inductor L_1 represents the outer conductor 412 of the outer coaxial cable unit 404 and the inductor L_2 represents the outer conductor 408 of the inner coaxial cable unit 402. The inductor L_3 represents the inner conductor of 410 of the outer coaxial cable unit 404 and the inductor L_4 represents the inner conductor 406 of the inner coaxial cable unit 402.

The antennas and antennas arrays of the present invention may also be used in a quadrature arrangement. For example, in Fig. 22, a quadrature antenna system 450 is shown comprising the hand, wrist and toe antenna array 100 of Figs. 9-12 and a pair of rectangular coaxial cable units 452a, 452b. Fig. 23 is a top view of the antenna arrays 100 and 452, showing the rectangular coaxial cable unit 452. In Fig. 23, the rectangular coaxial cable coil 452 lies in a flat plane A of the page. In Fig. 22, the flat plane A is perpendicular to the page and is perpendicular to the parallel planes P_1, P_2, P_3, of the antenna array 100.

In Fig. 22, the antenna array 100 and the rectangular or coaxial cable 452 are separated by a dielectric material 454, such as plastic. The dielectric material 454 may be in the form of a plate. The rectangular coaxial cable 452 may also be supported by a plate 456 of plastic or other dielectric material. The plates 502, 503 may be 1 inch thick, for example.

Returning to Fig. 23, the rectangular coaxial cable unit 452 preferably includes two sections 452a and 452b to lower the inductance of the coaxial cable 452. The two sections 452a, 452b each have an inner conductor 12', 12", and an outer conductor 14', 14", as in Fig. 7. Opposing ends 16a, 18a and 16b, 18b of the inner conductors 12' and 12" are connected to each other through capacitors C_1 and C_4, while the opposing ends 20a, 22a and 20b, 22b of the outer conductors 14' 14", are connected to each other through capacitors C_2, C_3. The output of the rectangular coaxial cable unit 452 is across the capacitor C_1. As above, the capacitor C_1 may be connected to a varactor in the preamplifier of the MRI system or may be the varactor in the preamplifier.

A capacitor C_5 is connected across the output 108 of the antenna array 100 (see Fig. 9). The details of the connections of the inner and outer conductors of each coaxial cable unit of the array 100, and of the output across the capacitor C_5, are not shown in Fig. 23. As

above, the capacitor C_5 may also be connected to or be a varactor in the preamplifier of the MRI system.

The range of the rectangular coaxial cable units 452a, 452b may be extended by providing additional coaxial cable units 454, 456, perpendicular to the coaxial cable units 452a, 452b. Fig. 24 shows a quadrature antenna system 450' of Fig. 23, which includes additional coaxial cable units 454, 456. The coaxial cable unit 454 includes an inner conductor 458 and an outer conductor 460. The coaxial cable unit 456 includes an inner conductor 462 and an outer conductor 464. Fig. 25 is a side view of the quadrature antenna system 450, along line 25 in Fig. 26. Fig. 26 is a side view of the quadrature antenna system 450, along arrow 26 in Fig. 26. The height "h" of the coaxial cable units 454, 456 may be 4 inches, for example. The antenna array 100 is not shown in Figs. 25 and 26.

As shown in Figs. 24 and 25, a first end 460a of the outer conductor 460 of the coaxial cable unit 454 is connected to the second end 22b of the inner conductor 12" of the coaxial cable unit 452b through a capacitor C_{10}. The first end 458a of the inner conductor 458 is directly connected to the first end 18b of the outer conductor 14".

The second end 460b of the outer conductor 460 is directly connected to the second end 20b of the coaxial cable unit 452a. The second end 16b of the inner conductor 12' of the coaxial cable unit 452a and the second end 458b of the inner conductor 458 of the coaxial cable unit 454 are connected across a capacitor C_{11}.

As shown in Figs. 24 and 26, a first end 462a of the inner conductor 462 of the coaxial cable unit 456 is connected to the first end 18a of the inner conductor 12" of the coaxial cable unit 452b through a capacitor C_{12}. The second output of the quadrature antenna system is

provided across the capacitor C_{12}. The first end 464 of the outer conductor 464a is connected to the second end 22a of the outer conductor 14" through a capacitor C_{14}.

The second end 462b of the inner conductor 462 is directly electrically connected to the first end 16a of the inner conductor 12' of the coaxial cable unit 452a. The second end 464b of the outer conductor 464 is directly connected to the first end 20a of the outer conductor 14'.

The connections within the antenna array 100 are the same as in Fig. 23 and Fig. 9.

Fig. 27 is a schematic diagram of the preamplifier section 470 of an MRI system for use with the quadrature antenna systems 100 and 452. Two preamplifiers 472, 474 are shown, each providing an input 476, 478 to a phase shifter signal combiner 480, which provides an output 482 to the signal processing portion of an MRI system. The output from the antenna array 100 may be provided to the preamplifier 472 and the output from the rectangular coaxial cable antenna 452 may be provided to the preamplifier 472 for example. The phase shifter may be a Mini-Circuit 15542 ZSCQ-2-50, available from Mini-Circuit, Brooklyn, NY, for example.

Fig. 28 is a side view of the gap 490 between the magnet poles 492, 494 of an MRI system, such as the Quad 7000 available from FONAR, Corporation, Melville, NY. The quadrature antenna system 450 rests on a patient bed 496 portion of which is above the pole 494. Preferably, the quadrature antenna system 450 is supported at an angle with respect to the patient bed 496. An angle of about 15° may be provided, for example. The quadrature antenna system 450 may be supported at an angle by a wedge 498, for example.

The head and neck antenna array 200 may also be used in a quadrature arrangement by adding a coaxial cable unit perpendicular to the planes of the coaxial cable units of the antenna array 200, above or below the antenna array.

It is known that the voltage induced in a secondary winding of a transformer may be increased by increasing the number of turns in the secondary winding. The voltage induced in the inner conductor of the coaxial cable unit 10 of Fig. 1 may therefore be increased by connecting a plurality of inner conductors in series. The signal-to-noise (S/N) ratio is thereby improved. However, it is also known that connecting inductors in series increases the resistance of the circuit, decreasing the Q of the circuit. In addition, the inductance is also increased, preventing tuning of the circuit to high frequencies for use with magnets of high field strengths, as discussed above. Therefore, it is preferred to also provide inner conductors connected in parallel, to lower the resistance and inductance of the circuit including the inner conductors. Parallel connection also increases current flow, which also improves the signal-to-noise (S/N) ratio.

Fig. 29 is a plan view of a coaxial cable unit 500 in accordance with another embodiment of the invention, including multiple inner conductors. In the configuration of Fig. 29, four inner conductors 504, 506, 508, 510 are shown. More or fewer inner conductors may be provided. Fig. 29a is a cross-sectional view of the coaxial unit 500 along line 29a-29a in Fig. 29. Preferably, the inner conductors 504, 506, 508, 510 are closely and tightly bundled by slightly twisting them together. The inner conductors may be solid and are covered by a dielectric material, such as Teflon tubing (not shown).

The ends 502a, 502b of the outer conductor are connected across a capacitor C_1, as shown in Fig. 29.

A preferred connection scheme for the inner conductors is shown in Fig. 30, where the inductors L_2, L_3, L_4 and L_5 correspond to the inner conductors, 504, 506, 508, 510, respectively. The inductors L_2 and L_3 are preferably connected in series through capacitor C_2. The inductor L_3 and the inductor L_4 are also preferably connected in series, through a capacitor C_3. The inductors L_4 and L_5 are preferably connected in parallel. The parallel connector inductors L_4, L_5 are electrically connected to the inductor L_3 though a capacitor C_3. The resistance and inductance are not, therefore, increased as much as if all of the inductors were connected in series.

The circuit including the outer conductor and the circuit including the inner conductor are tuned to the same Larmor frequency, as discussed, above. As above, the capacitor C_4 may be connected to a varactor in the preamplifier of the MRI system or may be the varactor.

An antenna array may also be provided with multiple coaxial cable units 500. Fig. 31 shows three coaxial cable units 520, 522 and 524, separated by an insulator 526 such as Teflon®. Each coaxial cable 520, 522, 524, may be copper tubing with a diameter of about 0.25 inches. The Teflon® may be 1 mm thick, for example. Fig. 33 shows the three coaxial cable units 520, 522, 524 and the connections between their inner and outer conductors. Those connections will be explained with respect to the electrical schematic diagram of Fig. 33, wherein the top box corresponds to the coaxial cable unit 520, the middle box corresponds to the coaxial cable unit 522 and the lower box corresponds to the coaxial cable unit 524.

The outer conductors of each coaxial cable unit 520, 522, 524 are represented as inductors L_1, L_2 and L_3, respectively, in Fig. 33. The inductors L_1, L_2 and L_3 are preferably connected in parallel across a capacitor C_1. The parallel connection of the outer conductors

provides better shielding of the inner conductors. Alternatively, the outer conductors may be connected in series.

In the coaxial cable unit 520, two of the inner conductors are preferably connected in series and two are preferably connected in parallel. In Fig. 33, the inductors L_4, L_5, L_6 and L_7 represent the four inner conductors of the first coaxial cable unit 520. The two inductors L_4 and L_5 are connected in series across a capacitor C_2. The two inductors L_6 and L_7 are connected in parallel. The parallel connected inductors L_6, L_7 are connected to the inductor L_5 through a capacitor C_3.

In the second coaxial cable unit 522, two pairs of the inner conductors are connected in parallel. In Fig. 33, the inner conductors are represented by the inductors L_8, L_9, L_{10} and L_{11}, respectively. The inductors L_8 and L_9 are connected in parallel and the inductors L_{10} and L_{11} are connected in parallel. One pair (L_8, L_9) is connected to the other pair (L_{10}, L_{11}) across a capacitor C_4. The pair of parallel connected inductors (L_6, L_7) in the first coaxial cable unit is also connected to one of the pairs of parallel connected inductors (L_8, L_9) in the second coaxial cable unit 522 through a capacitor C_5.

In the third coaxial cable unit 524, three of the inner conductors are preferably connected in parallel. In Fig. 33, the inner conductors are represented as the inductors L_{12}, L_{13}, L_{14}, and L_{15}, respectively. The inductor L_{12} is connected to the three parallel connected inductors L_{13}, L_{14}, L_{15} through a capacitor C_6. While the capacitor C_6 is shown within the box 524 representing the third coaxial cable unit 524, it is most readily provided between the second and third coaxial cable units 522, 524, as shown in Fig. 32. One of the pairs of parallel connected inductors (L_{10}, L_{11}) in the second coaxial cable unit 522 is also connected in parallel to a first one of the inductors L_{12} in the third coaxial cable unit 524.

One end the first inductor L_4 of the first coaxial cable unit 520 and one end of the three parallel connected inductors L_{13}, L_{14}, L_{15} in the third coaxial cable unit 524 are connected across a capacitor C_7 to provide an output of the antenna array. As discussed above, the capacitor C_7 may be connected to a varactor in the preamplifier of the MRI system or may be the varactor in the preamplifier.

As above, the circuit of the connected inner conductors is tuned to the same frequency as the circuit including the outer conductors. Such a configuration was found to have a signal-to-noise (S/N) ratio of about 700 and a Q of about 200.

The outer conductor 504 of the coaxial cable unit 502 may have an inner diameter "D1" of 0.25 inches, as shown in Fig. 29a. The outer diameter "D2" (Fig. 29) of the coaxial cable unit ring may be 6 ¾ inches, for example. Such a coaxial cable unit is appropriate for imaging a knee, for example. The inner conductors 504, 506, 508 and 510 may each have an outer diameter of 0.74 millimeters. As stated above, the inner conductors 504, 506, 508, 510 are preferably tightly bundled.

For coaxial cable units with diameters D1 (Fig. 29) of greater than about 8½ inches, or for use with high magnetic field strengths and higher Larmor frequencies, it is preferred to connect all of the inner conductors in parallel, to facilitate tuning. A parallel connection scheme is discussed below with respect to Fig. 36.

In another embodiment, an additional outer conductor may be added to the coaxial cable unit 500 of Fig. 29, forming a triaxial cable unit 600, as shown in Fig. 34. Two outer conductors 602, 604 are provided with multiple inner conductors 606, 608, 610, 612. The outermost, first outer conductor 602, which surrounds the second outer conductor 604, has

openings 614 distributed around its periphery. Fig. 34a is a cross-sectional view of the triaxial cable unit 600 along line 34a-34a of Fig. 34, showing the various components.

Ends of the first outer conductor 602 are connected through a capacitor C_2. Ends of the second outer conductor 604 are also connected to each other through a capacitor C_1. The first and second outer conductors 602, 604 are both tuned to the Larmor frequency of the species of interest. Depending on the diameter D of the ring 600, and/or the magnetic field strength, the inner conductors 606, 608, 610, 612 are connected to each other as described above with respect to Fig. 30, or in parallel, as discussed below.

Fig. 35 is a schematic representation of a circuit corresponding to the triaxial cable unit of Fig. 34, wherein the first outer conductor 602 is represented by the inductor L_1, the second outer conductor 604 is represented as the inductor L_2 and the inner conductors 606, 608, 610, 612 are represented as inductors L_3, L_4, L_5, L_6, respectively. As discussed above, ends of the outer conductors 602, 604 (L_1, L_2) are connected across the capacitors C_1, C_2, respectively. Two of the inner conductors 606, 608 (L_3, L_4) are connected in series and the other two inner conductors 610, 612 (L_5, L_6) are connected in parallel, as shown in Fig. 35.

Fig. 36 is a schematic representation of the triaxial cable unit 600 of Fig. 34, wherein the inner conductors 606, 608, 610, 612 (L_3, L_4, L_5, L_6) are connected in parallel.

In one configuration, the triaxial cable unit ring has an inner diameter "D" of 8 inches. The inner diameter "d1" (Fig. 34a) of the first conductor is 3/8 inch. The inner diameter "d2" of the second conductor is ¼ inch. The holes 614 in the first conductor 602 have diameters of about 2 mm. and are separated by a length "L" of about 1 inch.

Because of the holes 614 through the first, outermost conduct 602, it is believed, without limiting the scope of the invention, that magnetic resonance signals are detected by both

the first and second outer conductors 602 and 604, and that in addition, since the first and second outer conductors 602, 604 are inductively coupled, voltage signals which are a function of the magnetic resonance signals detected by the first outer conductor 602, are induced in the second outer conductor 604. The second outer conductor is also inductively coupled to the inner conductors 606, 608, 610, 612 and induces voltage signals in the inner conductors which are provided to the MRI system for processing. While the embodiment of Figs. 34-36 has multiple inner conductors 504, 506, 508, 510, a second outer conductor 604 may be used in a triaxial cable with only one inner conductor.

As mentioned above, the coaxial cable units of the present invention may also be used as transmitting antennas. Fig. 37 shows a coaxial cable unit 700 as in Fig. 3, wherein the ends 16, 18 of the inner conductor 12 are connected across a capacitor C_1, to a radio frequency power source. The ends 20, 22 of the outer conductor 14 are connected across a capacitor C_2. The inductance of the inner and outer conductors 12, 14 and the values of the capacitors are adjusted so that the respective circuits are tuned to substantially the same frequency.

Fig. 38 is a schematic representation of the coaxial cable unit of Fig. 37, where the inner conductor 12 is represented by the inductor L_1 and the outer conductor 14 is represented by L_2. In this case, a time varying voltage in the inner conductor 12 (L_1) induces a time-varying voltage in the outer conductor 14 (L_1), causing emission of radio frequency pulses from the outer conductor 14 (L_1). The first circuit A comprising the inductor L_1 and the capacitor C_1 filters the driving signal from the RF power source to within a first bandwidth. The second circuit B comprising the inductor L_2 and the capacitor C_2 filters the signal received from the first circuit A to within a narrower bandwidth. The signal transmitted by the circuit B therefore includes less noise than conventional transmitting antennas.

Any of the coaxial cable configurations discussed above can act as both a receiving and transmitting antenna.

Fig. 39 is a schematic representation of portions of an MRI system 800, showing in particular a connection between an antenna or antenna array 802 in accordance with the present invention with certain components of the MRI system. A signal processing system 804, a computer 806, an NMR controller 808, a gradient coil subsystem 810, a receiver subsystem 812 including a preamplifier 116, and an image display system 114 of the MRI system 800 are also shown. The computer 806 controls the overall operation of the MRI system 800. The NMR controller 808 stores the timing of the scanning sequences and controls implementation of the scanning sequence. The signal processing system 804 typically includes a variable amplifier, a frequency down converter, an analog-to-digital converting array and a digital data processor (not shown), as is known in the art. The gradient coil subsystem 810 includes the gradient coils and a gradient waveform generator, which outputs particular waveforms for a desired scanning sequence to the gradient coils under the control of the NMR controller 808, also as is known in the art. The particular ends of the inner conductor of the coaxial cable unit providing the output of the antenna array 802 are connected to the receiver system 812 of the MRI system 800 through the port 818.

As discussed above, the presence of a patient provides a load on the antenna array 802 which lowers the antenna's Q. The presence of the patient also shifts the resonant frequency of the antenna array 802, which may require retuning to the desired Larmor frequency. A varactor, or variable capacitor 820 is therefore provided between the capacitor C_1 of the antenna 802 and the preamplifier 816 parallel to the capacitor C_1, to enable retuning of the antenna array 802 when the antenna array is positioned with respect to the patient, as is known in the art.

Alternatively, C_1 may be the varactor 520. A back diode 822 is preferably provided parallel to the varactor 820 to prevent the passage of excessive voltage to the preamplifier 816, also as is known in the art. Voltage greater than about 0.7 volts is typically blocked by the back diode 822. The varactor 820 is controlled by the computer 806. The port 818 may be connected to the varactor 820, back diode 552 and preamplifier 810 through a short, low capacitance cable, or other appropriate means.

Fig. 39 also shows an optional connection between the capacitor C_6 and the antenna 802, and the RF transmitting subsystem 824 of the MRI system, through the optional port 826. The RF transmitting subsystem includes an RF power source (not shown) for driving the transmitting antenna, as is known in the art. The RF transmitting subsystem 824 can also be connected to the capacitor C_1 through a switch controlled by the computer 806. The computer 806 would switch the connection between the RF subsystem 824 and the antenna 802, and the preamplifier 820 and the antenna 802, at appropriate times. The RF transmitting subsystem 824 is controlled by the NMR controller 808 and the computer 806 of the MRI system 800.

A more complete description of the structure and operation of the MRI system may be found in U.S. Patent No. 6,025,717, assigned to the assignee of the present invention and incorporated by reference, herein.

The above embodiments are examples of antennas and magnetic resonance imaging systems in accordance with the present invention. It will be recognized by those skilled in the art that changes may be introduced to those embodiments without going beyond the scope of the invention, which is defined by the claims, below.